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The effects of fall history on kinematic synergy during walking

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ABSTRACT

To prevent falls, control of the swing foot during walking is crucial. Recently, some studies demonstrated that the coordinated movement of lower limbs by kinematic synergy is important for stable walking. However, no study has been carried out to reveal the relation between falls and kinematic synergy, and it is unclear whether fall history alters the kinematic synergy. Thus, the purpose of this study was to test the effects of fall history on kinematic synergy using uncontrolled manifold (UCM) analysis. Older adults were divided into two groups: older adults without fall history (non-fallers, n = 14) and older adults with fall history of at least one fall in the 12 months prior to the measurements (fallers, n = 10). Subjects walked at their own comfortable speed on a pathway and kinematic data were collected. UCM analysis was performed to assess how variability of segmental configurations in the frontal plane, the mediolateral and vertical directions, affects the frontal trajectory of the swing foot. Fallers had a greater variability of segmental configurations than non-fallers in all phases. In the mediolateral direction, the kinematic synergy in fallers was significantly greater than that in non-fallers during the early and late swing phases. On the other hands, fallers continuously had greater kinematic synergy compared to non-fallers in the vertical direction. The results revealed that fall history increased the kinematic synergy, although fallers needed a greater variability of segmental configurations as a compensatory strategy to ensure kinematic synergy.

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1. Introduction

Falls in older adults are very common problems encountered in daily life, and it is important to maintain stability during walking (Imms et al., 1977). Falls often occur with lateral body motions, which are associated more with instabilities in the frontal plane as compared to those in the sagittal plane (Maki, 1997; Maki and McIlroy, 2006; Rogers and Mille, 2003). Indeed, previous studies have shown that sensitivity to perturbations during walking was much higher in the frontal plane than the sagittal plane, and walking requires active control by the central nervous system to maintain stability, particularly in the frontal plane (Bauby and Kuo, 2000; O'Connor and Kuo, 2009). To evaluate the risk of falling, several indices were used, such as walking speed, cadence, stride length, and step width (Maki, 1997; Barak et al., 2006). The swing foot trajectory is also important to create a proper base of support and to prevent falls (Rogers and Mille, 2003). Considering that swing foot is an end-effector of a multiple degree of freedom system during walking, uncontrolled manifold (UCM) analysis that calculates the segmental configurations related to swing foot trajectory is a feasible way to evaluate swing foot stability (Krishnan et al., 2013; Rosenblatt et al., 2015).

UCM hypothesis assumes that the central nervous system organizes kinematic variability at the level of multiple elemental variables (the multiple segmental angles in this study) to stabilize the performance variable (the swing foot trajectory in this study). The performance variable is potentially important to achieve coordinated and successful movements (e.g., the center of mass during standing, the swing foot during walking) (Scholz and Schöner, 1999). Recently, UCM analysis is applied to standing, hopping, and to upper limb movements (Kapur et al., 2010; Yen and Chang, 2010; Verrel et al., 2012; Hsu et al., 2014). UCM analysis calculates the variability of elemental variables across repeated trials, and is categorized into two components: variance within the UCM (V_{UCM}) and variance orthogonal to the UCM (V_{ORT}). V_{UCM} is the "good variability" that reflects stable performance variable, whereas V_{ORT} is the "bad variability" that reflects unstable performance variable. If V_{UCM} is greater than V_{ORT}, the segmental configuration is flexible and coordinated to stabilize a performance variable by kinematic synergy (Scholz and Schöner, 1999).

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Kinematic synergy is a proportional change in segmental configuration and contributes to reducing the number of hypothetical variables manipulated by the controller (Krishnan et al., 2013). High kinematic synergy implies high stability of a performance variable. Previous studies have shown that higher posture-stabilizing synergy index was seen in younger adults as compared to older adults during balance recovery after a quick perturbation (Hsu et al., 2013). On the other hand, Black et al. showed patients with Down syndrome had greater posture-stabilizing synergy during gait than healthy subjects (Black et al., 2007), suggesting the patients used compensatory strategy for successful gait. Older adults who had fall history might use the compensatory strategy due to their fall experiences. However, there is no study to investigate the effect of falls on kinematic synergy. Comparing the kinematic synergy between fallers and non-fallers is crucial to understand falls in older adults.

The purpose of the current study was to examine whether fall history affect the kinematic synergy in the frontal plane. The hypothesis was that the synergy stabilizing the swing foot in the frontal plane would be greater in fallers than in non-fallers through the swing phase.

2. Methodology

2.1. Participants

A total of 24 community-dwelling older adults participated in this study. The physical characteristics are shown in Table 1. Fall history was obtained by self-report of questionnaires and the subjects were divided into two groups: older adults without fall history (non-fallers, n = 14); and older adults who experienced at least one fall in the 12 months prior to the measurements (fallers, n = 10). Exclusion criteria for older adults included neurological disorders or musculoskeletal injuries that would affect performance, or an inability to walk without assistance. All subjects gave their informed consent to participate according to the procedures approved by the Research Ethics Committee of Kyoto University.

2.2. Protocol

Subjects walked a 6-m pathway at their own comfortable speed, repeating the walk 20 times after three practices. For steady state walking, we used forty steps except for the first four steps for further analysis (Halliday et al., 1998; Lindemann et al., 2008). We did not constrain walking speed, but we asked the participants to walk at a comfortable speed to evaluate their natural gait (Hausdorff et al., 2001). Twenty-two spherical reflective markers were placed on both sides of each participant at the following locations: first metatarsal, fifth metatarsal, medial malleolus, lateral malleolus, medial femoral condyle, lateral femoral condyle, medial tibia condyle, lateral tibia condyle, greater trochanter, and anterior superior iliac spine. The kinematics data were recorded with infrared cameras (VICON MX; Vicon Motion Systems, Oxford) at sample frequencies of 100 Hz. The marker data were used to identify the center of the joints in the frontal plane. The performance variable, the swing foot trajectory, was defined as the median between first and fifth metatarsal (TOE), and the position of the ankle, knee, and hip joint (femoral head center) was calculated with marker positions (Winter, 2009). Analysis was performed for the right limb from toe-off until heel contact because all participants were right-leg dominant. The swing data was normalized by time (0–100%). It was difficult to divide the swing data individually using the kinematic data; therefore, we divided the data equally into 0–33% (early-swing), 34–67% (mid-swing), and 68–100% (late-swing) periods (Perry, 1992).

Trajectories of the swing foot relative to the stance foot were measured in the mediolateral (ML) and vertical (V) directions, and the segmental angles were calculated (see the Section 2.3). The averages and the variabilities of the swing foot and the segmental angles were calculated across 40 steps. The variability of the swing foot was calculated as the standard deviation across the steps. For further statistical comparisons, the variables were averaged within three phases, early-, mid-, and late-swing. Additionally, general walking parameters, such as step length, step width, cadence, walking speed, and swing time as a percent of stride time, and stride length ratio as an index of gaits symmetry, were also measured to realize the features of both groups.

Table 1
Physical characteristics.

	Non-fallers	Fallers	*p* value
Age (years)	75.1 ± 5.4	78.0 ± 2.7	0.09
Height (m)	1.6 ± 0.09	1.6 ± 0.1	0.14
Weight (kg)	57.7 ± 7.8	53.0 ± 8.0	0.09
Step length (cm)	55.9 ± 5.0	51.60 ± 5.1	0.04*
Step width (cm)	9.8 ± 2.3	10.73 ± 2.5	0.33
Cadence (steps/min)	99.0 ± 8.2	96.83 ± 11.5	0.78
Walking speed (cm/s)	125.3 ± 15.6	101.42 ± 45.8	0.06
Swing time (%)	35.2 ± 3.6	36.7 ± 3.0	0.19
Stride length ratio	0.9 ± 0.05	0.9 ± 0.04	0.97

* $p \leq 0.05$ between non-fallers and fallers.

2.3. UCM analysis

UCM analysis was applied to evaluate the kinematic variability of multiple joints. A detailed description of UCM analysis and its application to the swing foot trajectory procedure are noted elsewhere (Krishnan et al., 2013; Scholz and Schöner, 1999). Krishnan et al. used a geometric model without bilateral feet and shank during the stance phase (Krishnan et al., 2013). In contrast, we created a new geometric model that included these segments, because the foot has important roles as a terminal joint in the kinetic chain for the lower limb (Donatelli, 1985), and the coordination of the stance limb that includes the shank is important (Papi et al., 2015). In brief, we created seven segments (pelvis, bilateral feet, shanks, and thighs) from each marker data, 14 degrees of freedom (DOF) geometric models, which correspond to segmental angles related to seven DOFs in the frontal plane and seven DOFs in the sagittal and transverse plane, to calculate the accurate length of segments (L) in the frontal plane, as shown in Fig. 1. These DOFs were chosen as elemental variables to stabilize a performance variable described as TOE in the current study. We applied the geometric mode to ML and V directions.

TOE trajectory in ML (TOE_{ML}) and V (TOE_V) directions and the elemental variable matrix were expressed as:

$$TOE_{ML} = L_1 \cos\alpha_1 \sin\theta_1 + L_2 \cos\alpha_2 \sin\theta_2 + L_3 \cos\alpha_3 \sin\theta_3 + L_4 \cos\alpha_4 \sin\theta_4 + L_5 \cos\alpha_5 \sin\theta_5 + L_6 \cos\alpha_6 \sin\theta_6 + L_7 \cos\alpha_7 \sin\theta_7$$

$$TOE_V = L_1 \cos\alpha_1 \cos\theta_1 + L_2 \cos\alpha_2 \cos\theta_2 + L_3 \cos\alpha_3 \cos\theta_3 + L_4 \cos\alpha_4 \cos\theta_4 + L_5 \cos\alpha_5 \cos\theta_5 + L_6 \cos\alpha_6 \cos\theta_6 + L_7 \cos\alpha_7 \cos\theta_7$$

The relationship between the TOE trajectory and elemental variable was estimated via the Jacobian (J). J is a matrix of partial derivatives of the performance variable with respect to each of the segmental angle variables. We used matrix decomposition to calculate the null-space of the J. The null space, ε, is (n - d) vectors; the number of dimensions in the segmental configuration space

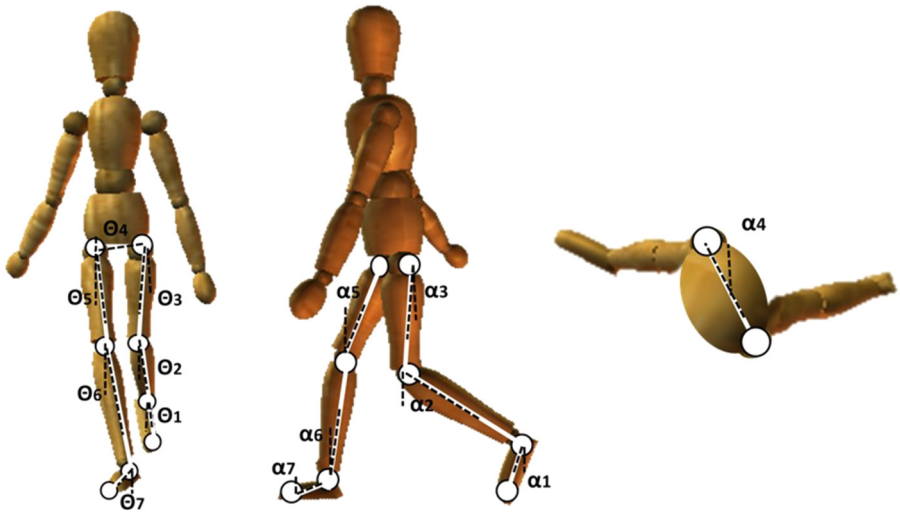

Fig. 1. Schematic illustrations of the segmental angles for the geometric model. Seven segments and 14 degrees of freedom, 7 degrees of freedom in the frontal plane (left foot, Θ_1: left shank, Θ_2: left thigh, Θ_3: pelvis, Θ_4: right thigh, Θ_5: right shank, Θ_6: right foot, Θ_7) and 7 degrees of freedom in sagittal and transverse plane (left foot, α_1: left shank, α_2: left thigh, α_3: pelvis, α_4: left thigh, α_5: right shank, α_6: right foot, α_7) were utilized for the analysis.

(n = 14) and the number of dimensions of TOE trajectories (d = 1) (Krishnan et al., 2013; Scholz and Schöner, 1999). At every percentage of swing, the differences of the segmental configurations from their mean $(\theta - \bar{\theta})$ were projected onto the null space:

$$\theta_{UCM} = \sum_{i=1}^{n-d} \left(\theta - \bar{\theta}\right) * \varepsilon_i$$

and onto a component orthogonal to this subspace:

$$\theta_{ORT} = \left(\theta - \bar{\theta}\right) - \theta_{UCM}$$

The variance (V_{UCM}) that does not affect the TOE trajectory was calculated as the between step average squared length of θ_{UCM}:

$$V_{UCM} = (n-d)^{-1} * N^{-1} * \sum (\theta_{UCM})^2$$

Similarly, the variance (V_{ORT}) that affects the TOE trajectory was calculated as:

$$V_{ORT} = d^{-1} * N^{-1} * \sum (\theta_{ORT})^2$$

Furthermore, the synergy index ΔV was calculated to evaluate the strength of the synergy from lower segments. This synergy index (Krishnan et al., 2013) was calculated as:

$$\Delta V = \frac{(V_{UCM} - V_{ORT})}{V_{TOT}}$$

where

$$V_{TOT} = \left(\frac{1}{n}\right)(dV_{ORT} + (n-d)V_{UCM})$$

The maximum ΔV is 14/13 (all variances lie within UCM), and the minimum ΔV is −14 (all variances lie in the orthogonal subspace); a more positive ΔV implies stronger synergy (Robert et al., 2009). For statistical analysis, we modified ΔV using Fisher's z-transformation (ΔV_Z) (Robert et al., 2009):

$$\Delta V_Z = 0.5 * \log\left[\frac{14 + \Delta V}{\left(\frac{14}{13}\right) - \Delta V}\right]$$

For the current model, $\Delta V_Z < 0.557$ implies the absence of synergy. To perform further statistical analyses, the segmental angles and indices of UCM analysis were averaged within the three phases: early-, mid-, and late-swing.

2.4. Statistical analysis

To evaluate the characteristics of the participants, independent t-tests were conducted to compare the age, weight, height, and general walking parameters between groups. Additionally, the average segmental angles and segmental angle variabilities were compared among the groups using independent t-tests in each phase. Also, analysis of variance for split-plot factorial designs were conducted to investigate the effects of *Group* (faller and non-faller) and *Phase* (early-swing, mid-swing, and late-swing) for the averaged swing foot, the swing foot variability, V_{UCM}, V_{ORT}, and ΔV_Z in ML and V directions, and the results were used to quantify whether fall history alters synergy index. When interaction effects were detected, post hoc comparisons were performed to test the differences in the variables between faller and non-faller or among early-, mid-, and late-swing. In cases of violations of normality, log-transformation was used prior to the parametric statistics. All statistical analyses were performed using SPSS software (PASW Statistics, Chicago) and the alpha level was set at 0.05.

3. Results

3.1. Physical characteristics

There was no significant difference was found for age, weight, height, and walking parameters except the step length ($p < 0.05$). We found that right shank angle (Θ_6) in early-swing was significantly lower in fallers ($p < 0.05$), although there was no significant difference in the other segmental angles. Regarding the segmental angle variabilities, variabilities of shank angle (α_6) during early- and mid-swing were significantly greater for fallers compared to non-fallers ($p < 0.05$). There was no significant effect of *Group* and *Phase* for averaged swing foot in V and ML directions. For swing foot variability, no significant effects of *Group* and *Phase* were found in ML direction, whereas we found a significant effect of *Group* ($F(1, 22) = 8.5$, $p < 0.01$) in V direction (Fig. 2). The swing

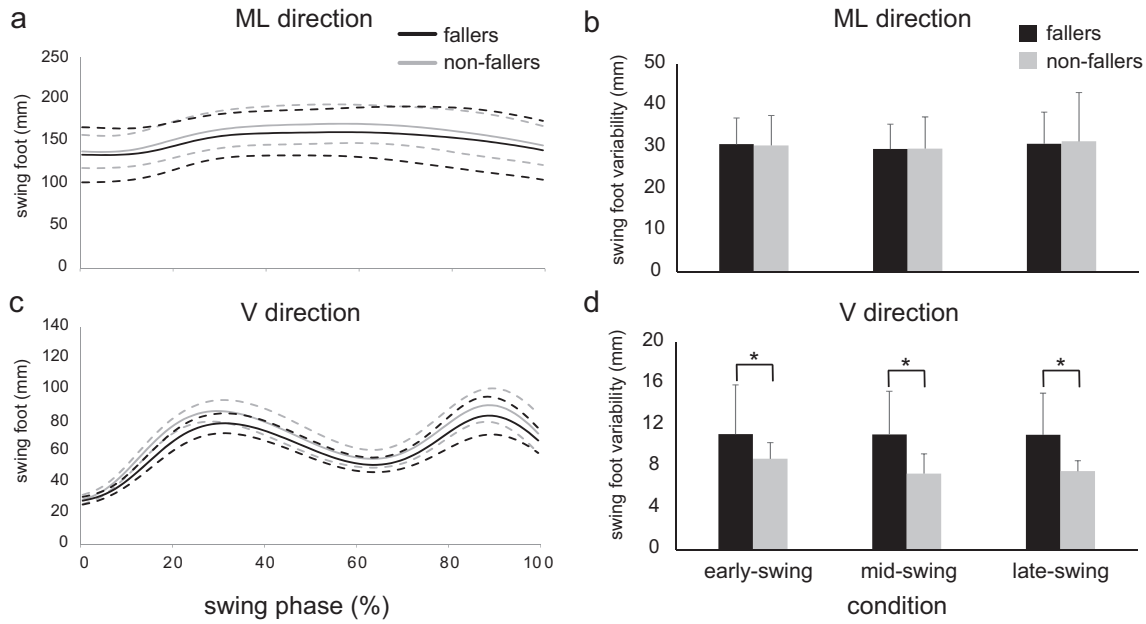

Fig. 2. The averaged time profiles of the swing foot (a and c) and the swing foot variability (b and d) across subjects with standard deviation. Left panels (a and c): the averaged time profiles of the swing foot in the normalized swing phases. The solid lines represent the averaged values, and the dotted lines represent the standard deviations; Right panels (b and d): the swing foot variability for three phases, early-swing, mid-swing, and late-swing. The error bars represent the standard deviations. The indices of the swing foot in fallers (black lines or black bars) and in non-fallers (grey lines and grey bars) are shown in mediolateral (ML) and vertical (V) direction. *Significant differences ($p < 0.05$) between fallers and non-fallers.

foot variability in fallers was significantly greater as compared to non-fallers in V direction. There were no interactions between factors for the swing foot variabilities.

3.2. Indices of UCM analysis

Fig. 3 shows the magnitudes of V_{UCM} in ML and V directions (Fig. 3a and b). For V_{UCM} in both directions, there were significant effects of *Group* (ML direction: $F(1, 22) = 140.3$, $p < 0.001$; V direction: $F(1, 22) = 140.1$, $p < 0.001$) and *Phase* (ML direction: $F(2, 44) = 27.1$, $p < 0.001$; V direction: $F(2, 44) = 22.2$, $p < 0.001$) with no interactions. V_{UCM} in fallers were significantly greater than those in non-fallers, and the V_{UCM} during late-swing was greater than that during early- and mid-swing, and V_{UCM} during mid-swing was lower than that during early- and late-swing in both ML and V directions.

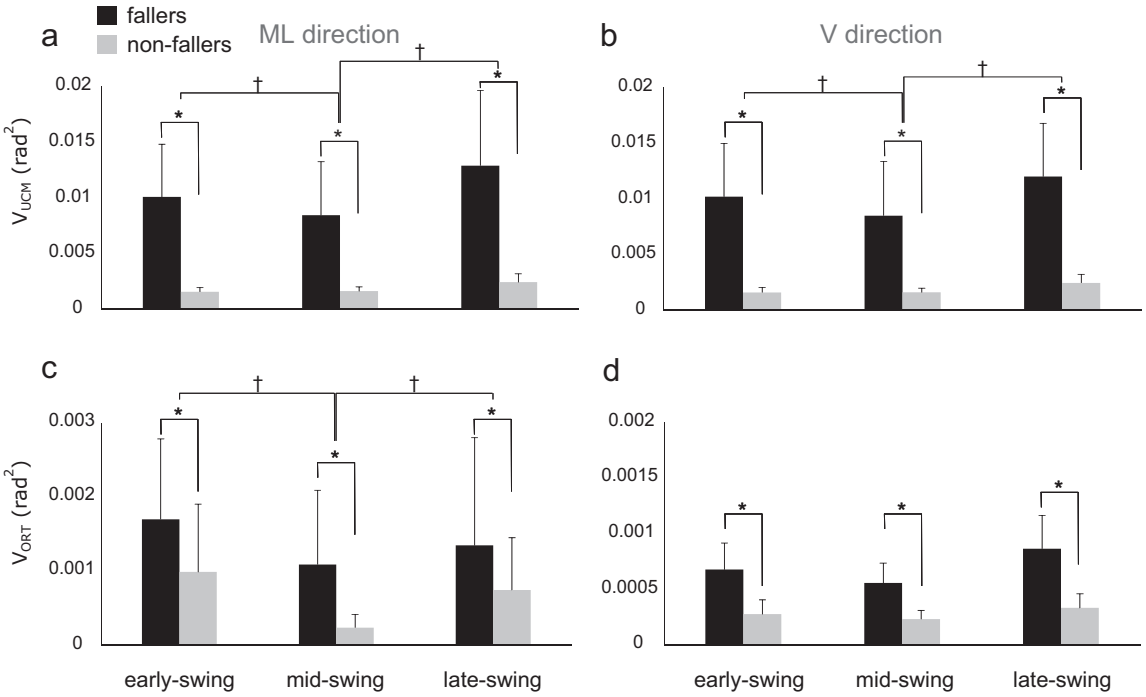

Fig. 3. Averaged variance components across subjects with standard deviation: V_{UCM} (a and b) and V_{ORT} (c and d) in fallers (black bars) and in non-fallers (grey bars) are shown. For abbreviation, see the caption for Fig. 2. *Significant differences ($p < 0.05$) between fallers and non-fallers; †Significant differences ($p < 0.05$) between phases.

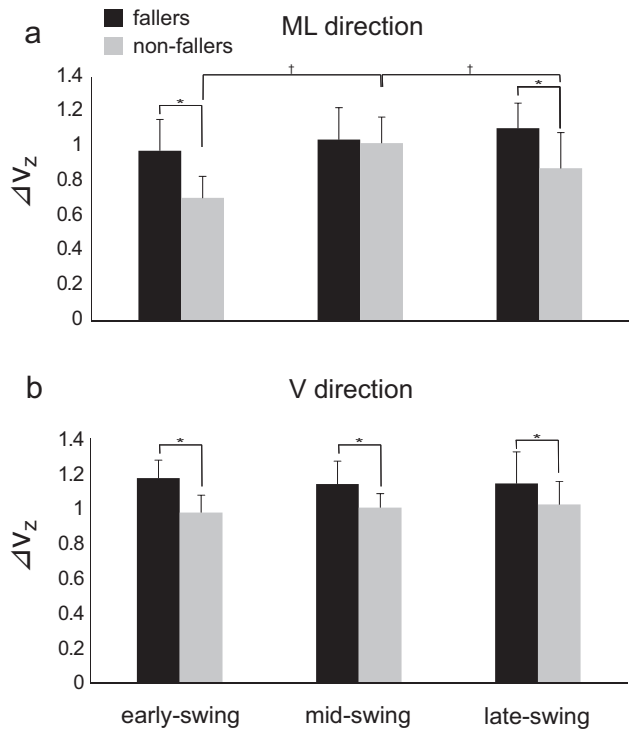

Fig. 4. Averaged synergy index across subjects with standard deviation: Synergy index (ΔV_Z) in fallers (black bars) and in non-fallers (grey bars) are shown in ML (a) and V (b) directions. For abbreviation, see the caption for Fig. 2. *Significant differences ($p < 0.05$) between fallers and non-fallers; †Significant differences ($p < 0.05$) between phases.

The magnitudes of V_{ORT} are presented in Fig. 3c and d. For V_{ORT} in ML direction, there were significant effects of *Group* (F (1, 22) = 9.4, $p < 0.01$), *Phase* (F (2, 44) = 53.5, $p < 0.001$), and *Group* × *Phase* (F (2, 44) = 11.4, $p < 0.01$) interactions. V_{ORT} in fallers was greater during early-swing as compared to mid- and late-swing, whereas V_{ORT} in non-fallers was lower during mid-swing as compared to early- and late-swing. Additionally, V_{ORT} in fallers were significantly greater than those in non-fallers in all phases. In V direction, we found the effects of *Group* (F (1, 22) = 78.8, $p < 0.001$), and *Phase* (F (2, 44) = 8.3, $p < 0.01$), with no interaction. V_{ORT} during late-swing was significantly greater as compared to the other phases. V_{ORT} in fallers was significantly greater than those in non-fallers, similar to V_{ORT} in ML direction.

The magnitudes of ΔV_Z are presented in Fig. 4. For ΔV_Z in both directions, the fallers and non-fallers were above 0.557 in all three phases, thus indicating the existence of synergy (see Methods). In ML direction, significant effects of *Group* (F (1, 22) = 9.0, $p < 0.01$), *Phase* (F (2, 44) = 21.3, $p < 0.001$), and *Group* x *Phase* interaction (F (2, 44) = 9.7, $p < 0.01$) were found. As results of the interaction, during early-swing and late-swing, ΔV_Z values in fallers were significantly greater than in non-fallers, although there was no difference during mid-swing. ΔV_Z in fallers was no significant difference across phases, whereas ΔV_Z in non-fallers was greater during mid-swing as compared to early- and late-swing. On the other hand, we found significant effect of *Group* (F (1, 22) = 16.2, $p < 0.01$) in V direction, with no interaction. ΔV_Z in fallers was significantly greater as compared to non-fallers.

4. Discussion

The purpose of this study was to investigate whether fall history alters the strength of the kinematic synergy. We found the significant effect of fall history on the synergy index (ΔV_Z) in ML and V directions. In ML direction, ΔV_Z in fallers was greater than that in non-fallers during early-swing and late-swing, while there was no significant difference during mid-swing. In V direction, ΔV_Z in fallers were greater than that in non-fallers throughout the swing phase. These observations were supported our hypothesis, that is, the synergy stabilizing the swing foot in the frontal plane would be greater in fallers than in non-fallers through the swing phase. Both variance components, V_{UCM} and V_{ORT}, in fallers were continuously greater compared to non-fallers throughout the swing phase in both directions. Moreover, V_{UCM} in both fallers and non-fallers had the similar effect of phase in both directions. Also, the both groups had the similar effect of phase for V_{ORT} in V direction. In ML direction, V_{ORT} in non-fallers dropped from early-swing to mid-swing and increased from mid-swing to late-swing. On the other hands, an increase in V_{ORT} in fallers was not seen from mid-swing to late-swing, although there was a drop from early-swing to mid-swing.

In the ML direction, ΔV_Z in non-fallers increased from toe-off to mid-swing and decreased from mid-swing to heel contact, whereas ΔV_Z values in fallers were consistently high during all three phases. As a result of this interaction, ΔV_Z in fallers and non-fallers were similarly high during mid-swing. During this phase, healthy adults stabilized the swing foot by a decrease in V_{ORT} and an increase in synergy, and the phenomena strengthened when walking on a narrow walkway, bringing the stance and swing legs closer (Krishnan et al., 2013; Rosenblatt et al., 2015). These findings were similar to our results for fallers and non-fallers. It is possible that a higher ΔV_Z during mid-swing might occur to prevent the collision of stance and swing legs during walking.

During early-swing and late-swing, fallers had greater ΔV_Z due to an increase in V_{UCM} compared to non-fallers. These results were inconsistent with the general idea that high V_{UCM} and ΔV_Z reflect better performance (Hsu et al., 2013), because older adults with fall histories would not be "better walkers". We concluded that fallers could not use a strategy of purposeful destabilization (Park et al., 2012; Wang et al., 2014). Earlier studies showed that a decrease in synergy prior to an external postural perturbation or self-triggered perturbation was important as a feed-forward adjustment (Kim et al., 2006; Krishnan et al., 2011). Similar adjustments in synergy were also seen during movement, and it was shown that this function is related to the instability of the movement (Singh and Latash, 2011; Park et al., 2012; Wang et al., 2014). The differences in ΔV_Z between fallers and non-fallers might be related to the ability of feed-forward adjustments.

A previous study has shown that patients with Down syndrome walked with higher V_{UCM} as well as higher positional variability (Black et al., 2007). The authors interpreted that patients exploited a larger workspace and variability to compensate for the inherent mechanical motor instability (e.g., poor postural control), and as a result, an increase in synergy was observed (Black et al., 2007). Different motor strategies with high variability would be important to retain the performance variable for people who have poor postural stability. High foot variabilities in the V direction (not in ML direction) and V_{UCM} might reflect a compensatory strategy to minimize postural instability.

Another interpretation of the results is the effects of the threat of falling. Rosenblatt et al. (2014) showed healthy younger adults increased V_{UCM} when precision of foot placement is required during walking (e.g., walking in two lines marked on the pathway). Moreover, they found that healthy younger adults greatly increased V_{UCM} when the threat of walking was added (e.g., walking on a narrow and high block) (Rosenblatt et al., 2014). People who had fall experiences feel threatened by walking, and these previous experiences sometimes change the walking strategy of the central nervous system (Tinetti et al., 1990; Brown et al., 2002; Maki and Whitelaw, 1993). We did not evaluate the threat

of falling for our subjects, so in the future, it will be necessary to clarify if this threat leads to greater V_{UCM} during walking.

Compared to non-fallers, fallers had higher variabilities of the right shank in the sagittal plane in early- and mid-swing. Until early mid-swing, the shank and thigh movements by knee flexors are important to create sufficient foot clearance during walking, and these adjustments contribute to a proper foot placement (Patla and Prentice, 1995; Perry, 1992; Rankin et al., 2014). Given that the shank segment is linked to foot segment, fallers might use high shank variabilities for effective adjustment of foot position. The high shank variabilities possibly caused the high foot variabilities in the V direction, but the fallers stabilized the average swing foot position similar to non-fallers. The fallers can thus exploit this variability to retain the swing foot close to the average position.

The clinical significance of this study was to verify changes in the structures and features of variability due to fall history. Traditionally, the variability was utilized as an index to evaluate the risk of falling, and an increase in the variability was usually assumed because of the caution of falls (Maki, 1997; Barak et al., 2006). The results of UCM analysis, however, revealed that higher variability in older adults with fall histories stabilized the swing foot and was a motor strategy to compensate for some mechanical motor instabilities. UCM analysis can evaluate how variability is controlled with respect to a performance variable, and the indices may be able to sensitively evaluate the potential risk of falling. Future research will be necessary to clarify if the UCM indices are related to the actual fall incidents.

There were some limitations in the current study. First, we did not measure the effects of upper body on kinematic synergy during walking. While the upper body might not have an impact on the foot position directly, it could have an important impact upon persons' stability during walking. Second, we did not obtain detailed information about falls. The swing foot trajectory is mostly related to tripping (Berg et al., 1997; Stephen N Robinovitch et al., 2013), but the fall experiences also included falling by collapse and loss of consciousness (Stephen N Robinovitch et al., 2013). We were not able to confirm that tripping was the cause of falls for fallers.

5. Conclusions

Overall, our study is the first to demonstrate an alteration of kinematic synergy due to fall history. We revealed that fluctuation in a synergy index in mediolateral direction and increase in a synergy index in vertical direction can be important strategies during walking. The alteration of fallers might be a compensatory strategy resulting from the threat of walking. The indices of uncontrolled manifold analysis can be useful tools to evaluate gait stability before and after intervention.

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Conflicts of interest

The authors have no conflicts of interest to disclose.

References

Barak, Y., Wagenaar, R.C.C., Holt, K.G.G., 2006. Gait characteristics of elderly people with a history of falls: a dynamic approach. Phys. Ther. 86, 1501–1510. https://doi.org/10.2522/ptj.20050387.

Bauby, C.E., Kuo, A.D., 2000. Active control of lateral balance in human walking. J. Biomech. 33, 1433–1440. https://doi.org/10.1016/S0021-9290(00)00101-9.

Berg, W.R., Alessio, H.M., Mills, E.M., Chen, T.O.N.G., 1997. Circumstances and consequences of falls in independent community-dwelling older adults. Age Ageing 26, 261–268.

Black, D.P., Smith, B.A., Wu, J., Ulrich, B.D., 2007. Uncontrolled manifold analysis of segmental angle variability during walking: preadolescents with and without Down syndrome. Exp. Brain Res. 183, 511–521. https://doi.org/10.1007/s00221-007-1066-1.

Brown, L.A., Gage, W.H., Polych, M.A., Sleik, R.J., Winder, T.R., 2002. Central set influences on gait Age-dependent effects of postural threat. Exp. Brain Res. 145, 286–296. https://doi.org/10.1007/s00221-002-1082-0.

Donatelli, R., 1985. Normal biomechanics of the foot and ankle. Orthop. Sport. Phys. Ther. 7, 91–95.

Halliday, S.E., Winter, D.A., Frank, J.S., Patla, A.E., Prince, F., 1998. The initiation of gait in young, elderly, and Parkinson's disease subjects. Gait Posture 8, 8–14. https://doi.org/10.1016/S0966-6362(98)00020-4.

Hausdorff, J.M., Ashkenazy, Y., Peng, C.K., Ivanov, P.C., Stanley, H.E., Goldberger, A.L., 2001. When human walking becomes random walking: fractal analysis and modeling of gait rhythm fluctuations. Phys. A Stat. Mech. Appl. 302, 138–147. https://doi.org/10.1016/S0378-4371(01)00460-5.

Hsu, W.-L., Chou, L.-S., Woollacott, M., 2013. Age-related changes in joint coordination during balance recovery. Age (Dordr) 35, 1299–1309. https://doi.org/10.1007/s11357-012-9422-x.

Hsu, W.L., Lin, K.H., Yang, R. Sen, Cheng, C.H., 2014. Use of motor abundance in old adults in the regulation of a narrow-based stance. Eur. J. Appl. Physiol. 114, 261–271. https://doi.org/10.1007/s00421-013-2768-7.

Imms, F.J., Exton-Smith, A.N., Johnson, A.L., 1977. Falls in the elderly related to postural imbalance. Br. Med. J. 1, 261–264. https://doi.org/10.1136/bmj.1.6056.261.

Kapur, S., Zatsiorsky, V.M., Latash, M.L., 2010. Age-related changes in the control of finger force vectors. J. Appl. Physiol. 109, 1827–1841. https://doi.org/10.1152/japplphysiol.00430.2010.

Kim, S.W., Shim, J.K., Zatsiorsky, V.M., Latash, M.L., 2006. Anticipatory adjustments of multi-finger synergies in preparation for self-triggered perturbations. Exp. Brain Res. 174, 604–612. https://doi.org/10.1007/s00221-006-0505-8.

Krishnan, V., Aruin, A.S., Latash, M.L., 2011. Two stages and three components of the postural preparation to action. Exp. Brain Res. 212, 47–63. https://doi.org/10.1007/s00221-011-2694-z.

Krishnan, V., Rosenblatt, N.J., Latash, M.L., Grabiner, M.D., 2013. The effects of age on stabilization of the mediolateral trajectory of the swing foot. Gait Posture 38, 923–928. https://doi.org/10.1016/j.gaitpost.2013.04.023.

Lindemann, U., Najafi, B., Zijlstra, W., Hauer, K., Muche, R., Becker, C., Aminian, K., 2008. Distance to achieve steady state walking speed in frail elderly persons. Gait Posture 27, 91–96. https://doi.org/10.1016/j.gaitpost.2007.02.005.

Maki, B.E., 1997. Gait changes in older adults: predictors of falls or indicators of fear? J. Am. Geriatr. Soc. 45, 313–320. https://doi.org/10.1111/j.1532-5415.1997.tb00946.x.

Maki, B.E., McIlroy, W.E., 2006. Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. Age Ageing 35 (Suppl. 2), 12–18. https://doi.org/10.1093/ageing/afl078.

Maki, B.E., Whitelaw, R.S., 1993. Influence of expectation and arousal on center-of-pressure responses to transient postural perturbations. J. Vestib. Res. 3, 25–39.

O'Connor, S.M., Kuo, A.D., 2009. Direction-dependent control of balance during walking and standing. J. Neurophysiol. 102, 1411–1419. https://doi.org/10.1152/jn.00131.2009.

Papi, E., Rowe, P.J., Pomeroy, V.M., 2015. Analysis of gait within the uncontrolled manifold hypothesis: stabilisation of the centre of mass during gait. J. Biomech. 48, 324–331. https://doi.org/10.1016/j.jbiomech.2014.11.024.

Park, J., Wu, Y.-H., Lewis, M.M., Huang, X., Latash, M.L., 2012. Changes in multifinger interaction and coordination in Parkinson's disease. J. Neurophysiol. 108, 915–924. https://doi.org/10.1152/jn.00043.2012.

Patla, A.E., Prentice, S.D., 1995. The role of active forces and intersegmental dynamics in the control of limb trajectory over obstacles during locomotion in humans. Exp. Brain Res. 106, 499–504. https://doi.org/10.1007/BF00231074.

Perry, J., 1992. Gait Analysis: Normal and Pathological Function. SLACK Inc, Thorofare, NJ.

Rankin, B.L., Buffo, S.K., Dean, J.C., 2014. A neuromechanical strategy for mediolateral foot placement in walking humans. J. Neurophysiol. 112, 374–383. https://doi.org/10.1152/jn.00138.2014.

Robert, T., Bennett, B.C., Russell, S.D., Zirker, C.A., Abel, M.F., 2009. Angular momentum synergies during walking. Exp. Brain Res. 197, 185–197. https://doi.org/10.1007/s00221-009-1904-4.

Rogers, M.W., Mille, M.-L., 2003. Lateral stability and falls in older people. Exerc. Sport Sci. Rev. 31, 182–187. https://doi.org/10.1097/00003677-200310000-00005.

Rosenblatt, N.J., Hurt, C.P., Latash, M.L., Grabiner, M.D., 2014. An apparent contradiction: increasing variability to achieve greater precision? Exp. Brain Res. 232, 403–413. https://doi.org/10.1007/s00221-013-3748-1.

Rosenblatt, N.J., Latash, M.L., Hurt, C.P., Grabiner, M.D., 2015. Challenging gait leads to stronger lower-limb kinematic synergies: the effects of walking within a more narrow pathway. Neurosci. Lett. 600, 110–114. https://doi.org/10.1016/j.neulet.2015.05.039.

Scholz, J.P., Schöner, G., 1999. The uncontrolled manifold concept: identifying control variables for a functional task. Exp. Brain Res. 126, 289–306. https://doi.org/10.1007/s002210050738.

Singh, T., Latash, M.L., 2011. Effects of muscle fatigue on multi-muscle synergies. Exp. Brain Res. 214, 335–350. https://doi.org/10.1007/s00221-011-2831-8.

Robinovitch, Stephen N., Feldman, Fabio, Yang, Yijian, Schonnop, Rebecca, Lueng, Pet Ming, Sarraf, Thiago, Sims-Gould, Joanie, Loughin, Marie, 2013. Video

capture of the circumstances of falls in elderly people residing in long-term care: an observational study. Lancet 381, 47–54. https://doi.org/10.1016/S0140-6736(12)61263-X.Video.

Tinetti, M.E., Richman, D., Powell, L., 1990. Falls efficacy as a measure of fear of falling. J. Gerontol. 45, 239–243.

Verrel, J., Lövdén, M., Lindenberger, U., 2012. Normal aging reduces motor synergies in manual pointing. Neurobiol. Aging 33, 201–210. https://doi.org/10.1016/j.neurobiolaging.2010.07.006.

Wang, Y., Watanabe, K., Asaka, T., Wan, F., 2014. Muscle synergies in preparation to a step made with and without obstacle. Eur. J. Appl. Physiol. 114, 2561–2569. https://doi.org/10.1007/s00421-014-2978-7.

Winter, D.A., 2009. Biomechanics and Motor Control of Human Movement. John Wiley & Sons, Hoboken.

Yen, J.T., Chang, Y.-H., 2010. Rate-dependent control strategies stabilize limb forces during human locomotion. J. R. Soc. Interf. 7, 801–810. https://doi.org/10.1098/rsif.2009.0296.